Vessel Pose Estimation for Obstacle Avoidance in Needle Steering Surgery using Multiple Forward Looking Sensors

Vani Virdyawan\textsuperscript{1} and Ferdinando Rodriguez y Baena\textsuperscript{1}

Abstract—During percutaneous interventions in the brain, puncturing a vessel can cause life threatening complications. To avoid such a risk, current research has been directed towards the development of steerable needles. However, there is a risk that vessels of a size which is close to or smaller than the resolution of commonly used preoperative imaging modalities (0.59 x 0.59 x 1 mm) would not be detected during procedure planning, with a consequent increase in risk to the patient. In this work, we present a novel ensemble of forward looking sensors based on laser Doppler flowmetry, which are embedded within a biologically inspired steerable needle to enable vessel detection during the insertion process. Four Doppler signals are used to classify the pose of a vessel in front of the advancing needle with a high degree of accuracy (2° and 0.1 mm RMS errors), where relative measurements between sensors are used to correct for ambiguity. By using a robotic assisted needle insertion process, and thus a precisely controlled insertion speed, we also demonstrate how the setup can be used to discriminate between tissue bulk motion and vessel motion. In doing so, we describe a sensing apparatus applicable to a variety of needle steering systems, with the potential to eliminate the risk of hemorrhage during percutaneous procedures.

I. INTRODUCTION

Percutaneous interventions, such as biopsy [1], deep brain stimulation (DBS) implantation [2], and epilepsy treatment [3], are commonly performed in minimally invasive brain surgery. Generally, such procedures offer several benefits for the patient due to reduced trauma and a faster recovery time [4]. However, puncture of a blood vessel during the insertion process can pose a life threatening complication for which a robust solution that works with all vessel sizes has not yet been found [5], [6], [7], [8], [9]. In a recent study about DBS implantation, for instance, up to 5% of patients undergoing this procedure had such a complication [10].

Before surgery, preoperative imaging data are commonly used to plan a vessel-free insertion path, however the data can be unreliable due to intra-operative brain shift [11] and limited imaging resolution. There are in fact smaller vessels (> 0.2 mm in diameter [12]) within the working volume of the brain that would be too small to be detected by common imaging modalities such as Magnetic Resonance Imaging (MRI) and Computer Tomography (CT) Scanning, where voxel sizes are typically 0.59 x 0.59 x 1 mm [13]. To avoid the risk of puncturing such vessels during surgery, investigations have been conducted on forward looking, laser-based sensors mounted on the tip of rigid needles. The use of optical fibers to deliver and receive laser signals allows such sensors to be small (down to 150 μm in diameter [14]) and thus appropriate for inclusion in surgical needles and, to date, successful vessel detection systems include Optical Coherence Tomography (OCT) [15], Coherence Gated Doppler (CGD) [14], fluorescence [16], Laser Doppler Flowmetry (LDF) [17] and photoacoustic imaging [18].

The insertion path of a rigid needle, however, is limited to a straight line. Consequently, if a vessel is detected in front of the tip of the needle, the procedure must be interrupted. To tackle this problem, research has broadened in the direction of needles that can be steered around curvilinear paths to avoid obstacles [19]. There have been many designs of robotic steerable needles to date: bevel-tip needles [20], concentric tube needles [21], tendon actuated needles [22], [23], and biologically inspired programmable bevel tip needles (PBNs) [24]. Many efforts have also been directed towards embedding sensors in these needle, for position tracking, force sensing, and shape estimation [20], [25]. Conversely, only a handful of studies to date has focused on forward looking sensors embedded within a needle steering system. One of these is the work of Ayvali et al. [26], where an OCT probe was embedded in an active needle steering system. However, the work only focuses on the feasibility of the probe to be used while the needle is in bending. The other is our own feasibility study on the use of an LDF probe embedded within a 2.5 mm diameter PBN, which showed promise [27].

LDF probes have a large off-axis detection range (up to a few mm [28]) that makes them suitable to be deployed in PBNs, which possess a sizable offset between the tip and working channels, where optical sensors could be embedded (Fig 1). The LDF sensor detects a vessel by measuring the Doppler shift effect in the light refracted by the moving blood cells flowing within it. Laser light is delivered into tissue through an illumination fiber, scattered inside the tissue, and collected by a collection fiber. Due to this scattering, it is challenging to determine the source of the signal, and thus determining the absolute perfusion value of blood flow is not possible [29]. Therefore, the perfusion measurement is a relative value that is calibrated with Brownian motion of latex particles in water, with an arbitrary unit (AU). An LDF monitoring system also records the movement of the tissue, and this movement needs to be avoided since the movement of large scattering particles with slow movement will be erroneously recorded as a high perfusion value [30].

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It has been shown that vessel pose (i.e. position and orientation) prediction can be achieved by using successive measurements combined with a look up table of the inverse perfusion value obtained with two LDF probes [27]. However, this method only works for a given pair of tissue and vessel properties. In this paper, we propose a novel detection strategy based on the relative measurements of four probes embedded within a PBN prototype with a diameter of 4 mm (Fig. 1). Using these relative measurements, we propose a method where vessel pose estimation can be achieved for any tissue optical properties and without reliance on the inverse function of the perfusion value. In addition, we hypothesize that, if the needle is inserted using a constant insertion speed, the background perfusion value affecting vessel measurements would be constant. We exploit this assumption to suggest an approach where vessel pose can be discriminated from bulk tissue movements.

This novel method for vessel pose classification based on multiple forward looking sensors is presented in two parts. In Section II, a commercial laser Doppler blood-flow monitoring system is characterized for a brain tissue phantom. The characterization includes static measurements, where the perfusion value is measured for different vessel sizes and tissue properties, and dynamic measurements, where the perfusion value is obtained while the PBN is moving forward with constant insertion speed. In Section III, an algorithm based on the relative measurements between sensors is presented, which enables the vessel pose to be classified with a known degree of uncertainty. Section IV discusses these results, while a summary of the main findings and future work are included in Section V.

II. LASER DOPPLER PROBE CHARACTERIZATION

A. Material and Methods

A commercially available, four-channel laser-Doppler tissue monitor (OxyFlo™ Pro XL (Oxford Optronix, Abingdon, UK)) with bare-fiber type probes (NX-BF/F (Oxford Optronix, Abingdon, UK)) was characterized. The laser has a wavelength of 785 nm, with a power < 0.5 mW at the probe tip. The probe diameter is 0.3 mm. Perfusion values were recorded using a LabJack U3-HV (LabJack Corporation, Colorado, USA) with a 150 Hz update rate.

Firstly, perfusion values for different flow velocities (5, 20, 40, 80 mm/s) and vessel phantom sizes (0.3 and 0.6 mm inner diameter) were measured to investigate the dynamic range of the equipment. A capillary tube with an inner diameter of 0.6 mm and an outer diameter of 0.9 mm was used as the vessel phantom. To change the inner diameter to 0.3 mm, a polyethylene tube was placed within the capillary tube. The vessel phantom was embedded at a gradient inside a gray matter phantom [27]. The gray matter phantom was made by adding 3 g/L titanium dioxide (TiO₂) into 4.5% by weight of gelatin ($\mu_s' \approx 0.75$ mm$^{-1}$ and absorption coefficient $\mu_a = 0.005$ mm$^{-1}$)[31]. 3.5% fat milk, diluted in water to obtain the 25% volume fraction that has a similar reduced scattering coefficient to blood ($1.8$ mm$^{-1}$ with $\mu_a \approx 0.0015$ mm$^{-1}$) [32], was used as a blood surrogate. The flow velocities were achieved using an automatic syringe pump (Graseby 3200, Graseby Medical Ltd. UK). All measurements were performed with one Doppler probe perfectly aligned (i.e. 0.00 mm off-axis position) with the vessel, 0.30 mm away from the probe tip.

Secondly, perfusion values for gray matter and white matter phantoms were compared, where the white matter phantom was made by adding 15 g/L TiO₂ ($\mu_s' \approx 3.8$ mm$^{-1}$ and $\mu_a = 0.005$ mm$^{-1}$) into gelatin [31]. The comparison was done for a 0.6 mm diameter vessel phantom at 20 mm/s flow velocity, with 0.00 mm off-axis position and the vessel placed at different depths from the probe tip (0.30 - 2.10 mm, with 0.30 mm increments) and at different off-axis positions (-2.00 - 2.00 mm, with 0.20 mm increments) and the depth fixed at 0.30 mm. The gelatin box was mounted on the top of a kinematic base (KB25/M, Thorlabs, New Jersey, USA), which could be assembled and disassembled with high repeatability (26.72 µrad) so that the position of the vessel phantom with respect to the probe could be calibrated only once, beforehand. The bottom of the kinematic base was mounted on a two degree-of-freedom (DOF) precision linear stage. In order to calibrate the position of the vessel, the sensor was moved until it touched the vessel and the position was recorded. Based on the calibration of the whole setup, repeatability of ±0.10 mm was achieved in measuring the "zero position" of the vessel with respect to the probe.

The third set of experiments was conducted to investigate the effect of different constant insertion speeds on perfusion values. Four insertion speeds were tested: 0.2 mm/s, 0.4 mm/s, 0.6 mm/s, and 1.0 mm/s, all within the gray matter phantom. During insertion, the perfusion values were recorded and compared to the perfusion value obtained while the probe was stationary. On completion of this characterization experiment, the feasibility of detecting the pose of a vessel placed in front of the needle, as it traveled at 0.20 mm/s towards it, was investigated.

B. Results

Since determining an absolute perfusion value is not possible, an LDF monitor measures relative perfusion value with an arbitrary unit (AU). If the concentration of moving particles is kept constant, the perfusion value of an LDF monitor increases linearly with the increase of flow velocity up to the dynamic limit of the instrument [33]. Fig. 2a shows...
that the dynamic limit for our setup is 20 mm/s for both vessel phantoms. This flow velocity is equal to the flow velocity of an arteriole [34].

For a given flow velocity and vessel diameter, Fig. 2b and Fig. 2c show that the higher the reduced scattering properties, the lower the perfusion value. At 0.30 mm axial depth and 0.00 mm off-axis position, for instance, the perfusion value in the gray matter phantom gives 5 times the perfusion value of the white matter phantom (1500 AU compared to 300 AU). Even though the perfusion value in the gray matter phantom is not always 5 times the perfusion value of the white matter phantom, results demonstrate that, in general, this value is always higher.

The stationary probe experiment in the absence of a vessel within the phantom produced a perfusion value of 20 AU. Conversely, as can be seen in Fig. 2d, this perfusion value increases by 200 AU for 0.2 mm/s insertion speed and 400 AU for 0.4 mm/s insertion speed. A high spike was always observed at the beginning of the insertion for all of the experiments. In Fig. 2d there is another spike at 25 s. However, since it represents a sudden increase of perfusion value and is short in duration (0.6 s), we could safely assume that it was not due to an approaching vessel, and could thus discard it. Mean and standard deviation for 0.2, 0.4, 0.6, and 1.0 mm/s insertion speeds are shown in Fig. 2e. Statistically significant differences are observed for all of these measurements (one-way ANOVA, F: 317.4, P value: <0.0001 with Post-Hoc Tukey test). The virtually constant offset values identified can be fit linearly ($R^2 = 0.96$) with $B_0 = 1382 \times v - 137.4$, where $B_0$ is the perfusion offset value and $v$ is the insertion speed. The feasibility of detecting a vessel using 0.2 mm/s insertion speed is shown in Fig. 2f. Starting from 140 s, perfusion increased up to 956 AU. The insertion was then manually stopped. The perfusion value dropped to 780 AU, showing that there is a vessel in front of the probe. The perfusion value drop was approximately 5 times the perfusion value offset for 0.2 mm/s.

### III. Vessel Pose Classification

Due to the high variability in perfusion values as a function of tissue and vessel parameters, it would be near impossible to ascertain depth and off-axis position of a vessel for a given perfusion value, without prior knowledge of these parameters. In addition, using a single probe, it would be impossible to estimate the pose of a vessel lying in front of the tip of the needle. In the case of a rigid needle, the only objective is to detect the presence of a vessel. With a steerable needle, which has the ability to steer around an obstacle, information with regards to the vessel pose is required in order for a suitable escape procedure to be executed. In this work, a laser Doppler probe was embedded in each of the four segments of a PBN. These serve two purposes: 1) detect the presence of a vessel in front of any one of the four segments; 2) determine the possible vessel pose by comparing the simultaneous measurements from all four laser Doppler probes.

**A. Materials and Methods**

The probes were embedded in the PBN working channels (Fig. 1), at different $x - y$ positions but on the same orthogonal plane, when all the segments are aligned. The following assumptions were made: 1) only one vessel is ever in view, 2) the vessel is located on a plane which is approximately perpendicular to the insertion axis, 3) the portion of the vessel is straight inside the detection range of the sensors. On this basis, the distance between the vessel and the laser Doppler probe can be described as in Fig. 3a. The axis of the vessel is defined as:

$$ax + by + c = 0$$

where $-c/a$ is the x-intercept of the line, and $-c/b$ is the y-intercept of the line. The distance from each probe to the axis of the vessel is

$$d_n = |ax_n + by_n + c| / \sqrt{a^2 + b^2}$$

Here, $n = 1, 2, 3, 4$, and $d_n$ is the distance of the vessel to the respective probe. Based on Fig. 2c, the closer the off-axis distance of the vessel to the probe, the higher the perfusion value measured.

To classify the vessel pose, the measurement values were then sorted from highest to lowest: $P_{v1}, P_{v2}, P_{v3}, P_{v4}, P_{v1}$ being the highest and $P_{v4}$ the lowest perfusion values. $P_{v1}$ relates to the distance $d_{v1}$ between the vessel and the respective probe. Consequently, $d_{v1}$ is the shortest distance and $d_{v4}$ the longest. Since there were four probes, the total number of classes is equal to the number of permutations of the set $Pr = \{P_1, P_2, P_3, P_4\}$, which is $4! = 24$ classes. Boundaries of vessel pose classes are calculated by solving:

$$d_1 = d_2, d_3 = d_4, d_1 = d_4, d_2 = d_3, d_2 = d_4,$$ and $d_3 = d_4$.

The boundaries are then given by

$$a = -(b(y_n + y_m) + 2c)/(x_n + x_m)$$

and

$$a = -(b(y_n - y_m)/(x_n - x_m)$$

with $n = 1, 2, 3, 4$, $m = 1, 2, 3, 4$, and $n \neq m$.

In the equation of a line, the gradient is $-\frac{a}{b} = \tan \theta$. By setting $b = 1$, class boundaries can be obtained for $-90^\circ < \theta < 90^\circ$. For $\theta = 90^\circ$, the value of $b$ is set to 0 and $a$ is set to 1. Fig. 3b shows the boundary between classes as a function of $\theta$ and $c$, with probe positions given in Table I.

Relative measurements between probes are used to define the area of $\theta$ and $c$ that could result in the same measurement output. Since four probes are used, there are $C_2^4 = 6$

<table>
<thead>
<tr>
<th>Probes</th>
<th>$x$</th>
<th>$y$</th>
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<tbody>
<tr>
<td>1</td>
<td>-0.67</td>
<td>1.15</td>
</tr>
<tr>
<td>2</td>
<td>-1.15</td>
<td>-0.67</td>
</tr>
<tr>
<td>3</td>
<td>0.67</td>
<td>-1.15</td>
</tr>
<tr>
<td>4</td>
<td>1.15</td>
<td>0.67</td>
</tr>
</tbody>
</table>
comparisons to be made. Each comparison produces an area

\[ A_k = \begin{cases} 
    d_{k1} \leq d_{k2}, & \text{if } P_{k1} > P_{k2} \\
    d_{k1} = d_{k2}, & \text{if } P_{k1} = P_{k2} \\
    d_{k1} \geq d_{k2}, & \text{if } P_{k1} < P_{k2} 
\end{cases} \quad (5) \]

where \( k = 1, 2, 3, 4, 5, 6 \), \( d_{k1} \) is the distance from the vessel to the first probe, with perfusion value \( P_{k1} \), and \( d_{k2} \) is the distance from the vessel to the second probe, with perfusion value \( P_{k2} \). Possible vessel poses \( A_v \) are given by

\[ A_v = \bigcap_{i=1}^{6} A_k \quad (6) \]

An ideal sensor can discriminate a very small difference in off-axis vessel position. However, sensors used here were not ideal and thus suffered from limited sensitivity. Consequently, we used a tolerance \( d_t \) in our comparisons so that \( |d_{k1} - d_{k2}| \leq d_t \). As evident in Fig. 2c, the higher the perfusion value the steeper the slope between two off-axis distances. This means that \( d_t \) is also a function of the perfusion value. Based on our characterization results (Section II-B), we used two ranges of \( d_t \) so that

\[ A_k = \begin{cases} 
    |d_{k1} - d_{k2}| \leq 0.6, & \text{if } P_{k1} = P_{k2}, P_{k1} < 100 \text{AU} \\
    |d_{k1} - d_{k2}| \leq 0.2, & \text{if } P_{k1} = P_{k2}, P_{k1} \geq 100 \text{AU} 
\end{cases} \quad (7) \]

A schematic and the experimental setup for vessel pose classification are shown in Fig. 3c and Fig. 3d, respectively.

The needle was mounted in a rotation mount (RSP05/M, Thorlabs, New Jersey, USA) with a resolution of 2°, attached onto a three DOF linear stage with needle coordinates \((x, y, z)\) placed at the tip of the needle. The three DOF linear stage was used to change the \( z \) position of the vessel and to make small adjustments during the “zeroing” calibration. As in Section II-A, the gelatin box was mounted onto a kinematic base. To produce variable \( c \) values, the two DOF linear stage was translated in the \( y_{st} \) direction. For \(-90^\circ < \theta < 90^\circ\), the translation in \( y_{st} \) for a given \( c \) is

\[ d_{st} = c \cos \theta \quad (8) \]

where \( d_{st} \) is the translation of the linear stage in \( y_{st} \) from the zero position. For \( \theta = 90^\circ \), \( c \) is equal to \( d_{st} \).

Experiments were conducted for two cases. To ascertain the feasibility of the algorithm, a first set of experiments was carried out at \( 0^\circ \), for five \( c \) positions (0.40, 0.20, 0.00, -0.20, -0.40 mm). For each position, the perfusion values were measured for the vessel phantom at \(-1.80 \text{ - } 0.00 \text{ mm} \) position, with 0.30 mm increments. Based on these results, the second set of experiments was performed for 24 random pairs of \( c \) and \( \theta \) values (Table II). For each pair, the vessel was located at \( z = -0.60 \) mm. For all of the classification experiments, the flow velocity rate was set at 20 mm/s, with a 0.6 mm internal diameter capillary tube inside the gray matter phantom. All measurements were repeated at least 8 times. The classification was done based on the statistically
significant difference between measurements of each probe at a given position.

To avoid cross talk between each probe, the measurements were conducted by activating each channel in turn, after an initial calibration of each probe separately, using a motility standard provided by the manufacturer. The perfusion value from each probe was recorded for 5 s. The average value of this measurement was used to compare measurements between the probes.

To investigate the sensitivity of this method to the assumption that the vessel is perpendicular to the insertion axis (i.e. the tilt angle is zero), additional characterization of the gray matter phantom was conducted. Using 20 mm/s flow velocity, the perfusion values in the range of -2.0 - 2.0 mm off-axis position were measured in 0.2 mm increments at 6 vessel depths (0.3, 0.6, 0.9, 1.2, 1.5, 1.8 mm). If a vessel is not perpendicular to the insertion axis, the depth from the vessel to each one of the probes may not be the same. The characterization results were thus used to approximate the measurements of each probe for a given tilt angle. The method to predict the vessel pose developed here was then used to check the maximum tilt angle that would still allow the vessel pose to be predicted correctly. The maximum tilt angle of the vessel was investigated for \( \theta \) values ranging from -80° to 90° in 10° increments and a \( c \) value ranging between -2.4 mm to 2.4 mm in 0.2 mm increments. Predictions were made at \( z = -0.60 \) mm.

In addition, we also performed a Monte Carlo simulation [35], [36] to investigate the effect of changing the optical properties (\( \mu'_s \) and \( \mu_a \)) of the tissue. The simulation was carried out for \( \mu'_s \) values of 0.55, 0.75, and 0.95 mm\(^{-1}\), and \( \mu_a \) values of 0.005 mm\(^{-1}\) and 0.05 mm\(^{-1}\) (gray matter \( \mu_a \) [37]), for 20 mm/s flow velocity. Additional simulations were then performed to investigate the effect of optical properties on the prediction area.

### B. Results

All of the data were checked for normality using the Shapiro-Wilk normality test, with \( p = 0.01 \). Comparisons of perfusion values between each probe were performed using a one-way Anova, with post-hoc Tukey test. As can be seen from Fig. 4a - Fig. 4e, a high perfusion value (> 120 AU [27]) started to be detected in one of the probes at \( z = -1.20 \) mm. Hence, the classification algorithm was executed based on measurements at \( z = -0.60 \) mm, which is midway between the first detected \( z \) position and the tip of the needle.

For \( \theta = 0^\circ \), in the first experiments all of the vessel angles were correctly predicted. There is one error in the estimate for \( c = 0.40 \) mm, where the difference between the actual \( c \) value with the predicted value was 0.06 mm. All possible vessel poses are drawn as a gray area in Fig. 4f - 4j. The predicted values for 24 random pairs of \( \theta \) and \( c \) have 1.6° and 0.37 mm RMS error respectively, where the RMS error for \( \theta \) is less than the resolution of the rotation mount. High errors in \( c \) are found mainly for poses where the angle approaches 90° (Table II) and further data analysis was conducted to check for the source of these errors. As described in Section III-A, to set \( c \), the stage was translated in the \( y_{st} \) direction by \( d_{st} \). To check for the sensitivity of \( d_{st} \) as a function of \( \theta \), we used the directional error

\[
\text{dir}_{\text{error}} = c_{\text{error}} \cos \theta
\]

The directional RMS error was found to be 0.1 mm, which is equal to the uncertainty of finding the zero position of the vessel.

### C. Sensitivity Analysis Results

Fig. 5a shows the maximum tilt angle that the vessel prediction algorithm can handle. The average value of the tilt angle is \( \pm 13^\circ \). The simulation shows good agreement with the experiment (Fig. 5b). All perfusion values were normalized against the perfusion value at 0.3 mm depth and 0.00 mm off-axis value. Fig. 5c shows the effect of optical properties on the perfusion value. The highest perfusion value corresponds to \( \mu'_s = 0.55 \text{ mm}^{-1} \) and \( \mu_a = 0.005 \text{ mm}^{-1} \), while the lowest perfusion value corresponds to \( \mu'_s = 0.95 \text{ mm}^{-1} \) and \( \mu_a = 0.05 \text{ mm}^{-1} \). The simulations were performed by setting different optical properties in front of each probe. The parameter variation was based on the optical properties that result in the highest and the lowest perfusion value. The results show that predictions remain valid up to a maximum error of 10° in \( \theta \) and of 0.5 mm in \( \text{dir}_{\text{error}} \).

### IV. DISCUSSION

Static and dynamic characterization of a laser Doppler probe have been investigated in this report. Based on the static experiments, the laser Doppler monitor used here has a maximum dynamic range of 20 mm/s for both vessel sizes. However, an artery with a diameter of \( \approx 1 \) mm could have a flow velocity of up to 150 mm/s [38], which would be much higher than the dynamic limit of the commercial laser Doppler monitor used here. The development of a laser Doppler monitor that has higher dynamic range will be investigated in future work.

The tissue’s bulk movement is considered to be a significant problem in the use of laser Doppler systems [14], [30], [28]. Since tissue is also a scattering object, the movement of the tissue is measured by an LDF system that generates a higher perfusion value than it should. To avoid this, Wardell et al. [17] measured the perfusion along the DBS implantation path in an incremental manner. Our characterization of perfusion values with different constant insertion speeds shows that there is an additional, approximately constant perfusion offset that scales up linearly as a function of insertion speed. The materials in front of the tip of the needle are displaced during the insertion process. The Particle Image Velocimetry (PIV) method has been used to investigate the displacement and to track material point trajectories [39]. We found that the material point trajectories in front of the needle are quasi-static during constant insertion speed, and as such, these result in a constant background offset. Changing the optical properties could change the background perfusion value due to the difference between the number of scattering
Fig. 3. a) Representation of a vessel (red line approximation) detected by four probes \( P_1, P_2, P_3, P_4 \), b) Boundary of the 24 classes as a function of two parameters, \( \theta \) and \( c \), c) A schematic diagram of vessel detection experiments with the needle coordinate system \((x, y, z)\) located at the tip of the needle. The needle can be rotated with respect to its axis \((\theta_{\text{rm}})\) to change the \( \theta \) value. Moving the sample box in \( y_{\text{st}} \) direction changes the \( c \) value. By moving the needle in the direction of \( x_{\text{st}} \), more than one insertion can be done in a sample. d) Experimental setup used for vessel classification. The vessel and gray matter phantom were placed on top of a 2 degree of freedom (DOF) linear stage that can be moved in the \( x_{\text{st}} \) and \( y_{\text{st}} \) directions. The PBN was mounted on a rotation mount fixed to a three DOF linear stage.

Table II

<table>
<thead>
<tr>
<th>( \theta ) (°)</th>
<th>-80</th>
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<th>50</th>
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<th>110</th>
<th>140</th>
<th>170</th>
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<tbody>
<tr>
<td>( c ) (mm)</td>
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<tr>
<td>( c_{\text{error}} ) (mm)</td>
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<tr>
<td>( d_{\text{error}} ) (mm)</td>
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<td>0.04</td>
<td>0.00</td>
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Fig. 4. Perfusion value of four probes during insertion of the needle at different depths, starting at \( z = -1.80 \) mm with the following values for \( c \): a) \( c = 0.40 \) mm, b) \( c = 0.20 \) mm, c) \( c = 0.00 \) mm, d) \( c = -0.20 \) mm, and e) \( c = -0.40 \) mm. Possible vessel poses predicted using relative measurement of four probes: f) \( c = 0.40 \) mm, g) \( c = 0.20 \) mm, h) \( c = 0.00 \) mm, i) \( c = -0.20 \) mm, and j) \( c = -0.40 \) mm. Predictions were made at \( z = -0.60 \) mm. Gray area: predicted vessel pose, red line: vessel pose, \( P_1 \): probe 1, \( P_2 \): probe 2, \( P_3 \): probe 3, \( P_4 \): probe 4, \( O \): tip of the needle.
particles and the depth penetration of the light. However, since during the course of the insertion, the distribution of material point trajectories in front of the needle is quasi-static, we believe that the background perfusion value would hold constant as well.

Fig. 2f shows that a vessel located in front of the probe can still be detected by observing a monotonic increase of the perfusion value. Using 0.2 mm/s insertion speed, for instance, it takes $\approx 7$ minutes for full insertion of a 75 - 80 mm DBS electrode. This result demonstrates that detecting a vessel in front of a continuously advancing needle using a robotic assisted needle insertion system, combined with LDF, is indeed possible.

Our novel vessel pose classification algorithm employing multiple laser Doppler probes embedded within in a PBN shows an angular RMS error of $1.6^\circ$ and translational RMS error of 0.1 mm. Both of these errors are within the repeatability of the experimental setup itself, which is encouraging. As can be seen in Fig. 2c, the closer the vessel is to the Doppler probe, the higher the perfusion value. This relationship still holds for any set of homogeneous tissue optical properties. Therefore, even though specific tissue optical properties and vessel properties generates different perfusion values, by exploiting the relative measurements between probes, the perfusion values are normalized and the algorithm will predict vessel pose robustly.

In the simulation, the algorithm still performs well with a change of $\mu_s'$ from 0.55 to 0.95 mm$^{-1}$ and $\mu_a$ from 0.005 to 0.05 mm$^{-1}$. However, if the discrepancy between the optical properties in front of each probe is high, the algorithm may fail to predict the vessel pose correctly. This eventuality is unlikely due to the close proximity between the four probes ($\approx 1.9$ mm), as the tissue optical properties are not likely to vary significantly within such a confined space.

It should be noted that the prediction area corresponds to the area where the axis of the vessel lies. The classification algorithm is based on the position of the laser Doppler probes and the number of probes used, which means an optimization of the numbers and the position of the probes can be performed, for instance to obtain better vessel pose discrimination when $c$ is close to 0.

The axial distance ($z$) of the vessel from the tip of the needle cannot be discriminated here, however, due to the short axial detection range of the probe, a "no-go" region can be set from the tip of the needle up to 1.5 mm behind the predicted area. This no-go region can be loaded into an obstacle map that can be used to generate a new path to avoid the detected vessel. The short axial detection range means that a short retraction might be required to properly avoid the vessel. However, this short retraction should be preferable to puncturing the vessel.

Oldfield et al. [39] showed that there is tissue deformation around the needle during the insertion process, both in front of it and at the tissue-needle interface. Based on this work and the work of Liang et al. [15], it seems probable that a blood vessel would also move with the surrounding brain tissue, but sensors measurements could be used to update the vessel position in real-time.

Here, an assumption that a vessel lies in a plane perpendicular to the insertion axis was used. In fact, our sensitivity analysis shows that the average tilt angle that our method can handle is $\pm 13^\circ$. As can be seen in Fig. 5a, the maximum tilt angle varies depending on the vessel pose. The variation depends on whether the perfusion value order (from the highest to the lowest) for the four probes, changes. If the tilt angle does not introduce a change, the vessel pose will be robustly predicted even though a high tilt angle is present. To solve for possible vessel poses which do not comply with this restriction, a continuous update of the vessel classification, while the needle is steered around the vessel, would be needed, which will be the focus of future work.

V. CONCLUSION AND FUTURE WORK

In this report, multiple forward looking sensors for vessel pose classification were used in a 4 mm prototype of a bio-inspired Programmable Bevel-Tip Needle. Due to the promising, sub-millimeter accuracy results, the method will be used in a smaller prototype of the bio-inspired PBN that has a diameter of 2.5 mm and 8 working channels along its length. Specifically, in future work, several aspects of the
laser Doppler monitor will be improved, i.e. by developing faster measurement equipment, smaller laser Doppler probes that would fit into the 2.5 mm diameter PBN, and a deeper axial detection range. In addition, integration of the method within a real-time path planner will be explored. The concept of multiple forward looking sensors in a steerable needle to classify vessel pose can be translated into other types of steerable needle systems, enabling safer insertion with obstacle avoidance even for vessels that cannot currently be detected with preoperative imaging.

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REFERENCES


[34] I. Fredriksson, Quantitative Laser Doppler Flowmetry, 2009, no. 1269.


